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Compensatory mechanisms of individuals with above-knee amputation in response to steady-state walking speed and mobility classification

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Abstract

Compensatory mechanisms of individuals with above-knee amputation in response to steady-state walking speed and mobility classification

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Individuals with above-knee amputations utilize compensatory mechanisms to walk due to the functional loss of muscles spanning the knee and ankle joints and the use of mechanically-passive prostheses. Clinical classifications of an individual's mobility (i.e. their k-level) are coarsely defined based on whether an individual can walk at multiple cadences and dependent or independent of a walking aide. However, it is unclear what biomechanical mechanisms they use to both walk and modulate their walking speed. The objective of this study was to investigate the kinematic and kinetic trajectories of the amputated and non-amputated limbs during walking at widely-varying speeds, as well as overall joint mechanical work computed over subjects' gait cycles in individuals with differing clinical classifications of mobility. We hypothesized individuals with a higher k-level would walk with more symmetry due to having better strength, control over their residual limb, and use of their prosthesis, and thus be less reliant on the contralateral leg to walk at faster speeds. Not only was our hypothesis not supported, our data analysis supports the opposite notion – that higher k-level individuals walk with greater asymmetry of joint

mechanics than lower k-level individuals. The development of active prostheses controllers has been a focus of rehabilitation research through the use of microprocessors and motors to restore the functional performance of individuals with amputation. The asymmetric joint trajectories used to walk with passive prostheses motivate a need to design adaptive protheses controllers than can be personalized for an individual’s baseline ability, especially at faster speeds.

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INTRODUCTION

Above-knee amputations (AKAs) have been performed on over 450,000 adults in the US between 1998 and 2013, due primarily to vascular diseases, trauma, periprosthetic joint infection, and malignancy [1]. Chronic pain is highly prevalent across persons with limb loss, irrespective of time since amputation [2]. Additionally, individuals with amputation tend to ambulate with asymmetry, favoring their contralateral, or non-amputated limb and increasing its loading, leading to degenerative changes in the contralateral limb, such as osteoarthritis of the hip joints. Conversely, spending less time on the ipsilateral, or amputated limb with insufficient loading of the bones leads to osteopenia and subsequent osteoporosis [3]. Numerous other secondary complications have been shown to arise including back pain, skin irritation, and general deconditioning due to prosthesis use [4]–[6]. Investigating the kinematic and kinetic causes of asymmetry in response to increasing steady state walking speed and the functional ability of the individual is essential for developments in clinical and research fields. Analysis deepens our understanding of the causes of secondary complications, aides in the design of active prostheses to counteract asymmetries, and improves the functional abilities of individuals with amputation.

Individuals with AKAs are reported to have slower walking speeds than healthy individuals [7], [8] . The prosthetic walking potential of individuals with lower limb amputation is done using a classification system referred to as ‘K-levels’ ranging from K0 (lowest mobility) to K4 (highest mobility) [9], [10]. This classification scale is coarsely defined based on whether an individual can walk at multiple cadences and dependent or independent of a walking aide. However, it is unclear what biomechanical mechanisms they use to both walk and modulate their walking speed. A primary factor of asymmetry and reduced forward propulsion capability is due to the loss of the knee extensors and ankle plantar flexors which are essential for body support, mechanical power, and leg swing control [11], [12]. Gait analysis has shown asymmetries in ground reaction forces, joint kinematics, and joint kinetics as a result of the loss of these joint spanning muscles [13]. Prostheses have been designed with reduced inertial properties to improve comfort, socket stability, and to lower metabolic expenditure [14], [15]. However, the physiological leg typically weighs more, leading to fundamental kinematic and kinetic differences [16].

Previous studies have indicated the contralateral leg produces larger GRFs that increase with walking speed relative to the ipsilateral leg [17]. Prosthetic foot-ankle, and knee joints in these studies were passive and thus limited in their contribution to forward propulsion, especially as walking speed increases. A commonly observed compensatory mechanism is for the ipsilateral hip to produce significantly more flexor work due to the lack of prosthetic knee and ankle flexion during the stance phase compared to the contralateral limb [18]. It has been shown that ipsilateral leg deficits in propulsion increase at higher walking speeds, however the asymmetries exhibited by individuals with different classifications of mobility has not been studied as closely.

The objective of this study was to investigate the kinematic and kinetic trajectories of the amputated and non-amputated limbs in the sagittal plane during walking at widely-varying speeds, as well as overall joint mechanical work computed over subjects' gait cycles in individuals with different clinical classifications of mobility. We hypothesized that, regardless of functional ability, individuals would be more reliant on their contralateral than ipsilateral limb to maintain steady-state walking speed, and this reliance would increase with walking speed. We also hypothesized individuals with a higher k-level, or functional ability, would walk with more symmetry due to having better strength, control over their residual limb, and use of their prosthesis, and thus be less reliant on their contralateral leg, while following the trend of our first hypotheses to increase reliance on their contralateral limb as walking speed increases.

METHODS

Participants and Data Collection

Experimental data published by the Bionic Engineering lab at the University of Utah consisted of 18 individuals with unilateral-above knee amputations recorded at five steady-state walking speeds based on their ability to comfortably walk [19]. The subjects were classified in two groups of nine, K2 and K3, walking at their comfortable range of speeds: 0.4 – 0.8 m/s and 0.6 – 1.4 m/s, respectively (additional subject information can be found in the appendix). Both the K2 and K3 group walked at the same steady-state speeds of 0.6 and 0.8 m/s. Subjects received amputation over one year prior to the experiment and self-reported using a passive prosthesis over three hours a day for over six months. Motion capture data (Vicon, Inc.) and GRFs from an instrumented treadmill (Bertec, Inc.) were used to generate kinematic and kinetic joint data, parsed from heel strike to heel strike.

Following a residual analysis, a low-pass Butterworth filter with a cut-off frequency of 6 Hz and 15 Hz was applied to the marker trajectory and force plate data, respectively [20]. The inertial properties of each segment, the subject's mass, and center of mass were determined using Dempster's and Hanavan's assumptions, reported weight, and segment dimension approximations [21], [22]. The knee joint center of rotation was found using Symmetrical Axis of Rotation Analysis (SARA) [23], and the hip joint center of rotation was found using the Charnwood Dynamics Model (CODA) [24], [25] . The prosthesis side shank weight was approximated to have 1/3 of the non-amputated shank weight, with the center of mass 25% below the top of the shank [26].

Data Analysis

Statistical parametric mapping (SPM) of joint moment and angle was used to find temporal differences between the ipsilateral and contralateral limbs, and the two groups when walking at the same speed [27]. SPM allows statistical comparisons over complex biomechanical trajectories so regions of significant difference instead of singular points can be investigated [28]. A repeated measures one-way ANOVA was conducted using the spm1d (spm1d.org) package in Matlab to generate ‘within-subject’ comparisons. A one-way ANOVA was conducted to generate ‘between-subject’ comparisons of the same limb of the K2 and K3 group when walking at the same speed. Positive and negative work of the hip, knee, and ankle were computed as the integral of the joint moment with respect to its angle, and compared between limbs and the two groups. A two-sample t-test was conducted with $\alpha = 0.5$ and a Bonferroni correction for multiple comparisons.

RESULTS

The results obtained from SPM of one gait cycle, from heel strike to heel strike, are illustrated in *Fig. 1* and *Fig. 2*. Average joint angle and moment trajectories of the 9 subjects in each group were plotted for each joint at 5 steady-state walking speeds. The K2 group and K3 group trajectories are in the left and right columns, respectively. The large light grey square near the top or middle of the plots designates zero joint angle or moment, with positive denoting joint extension and negative denoting joint flexion. Red indicates the ipsilateral, or amputated limb, with the intact hip and prosthesis joints for the ankle and knee. Blue indicates the contralateral, or non-amputated limb. Colors transition from light to dark from slower to faster walking speeds. The gray regions at the bottom of the plot indicate regions of significant difference between the ipsilateral and contralateral joints from SPM.

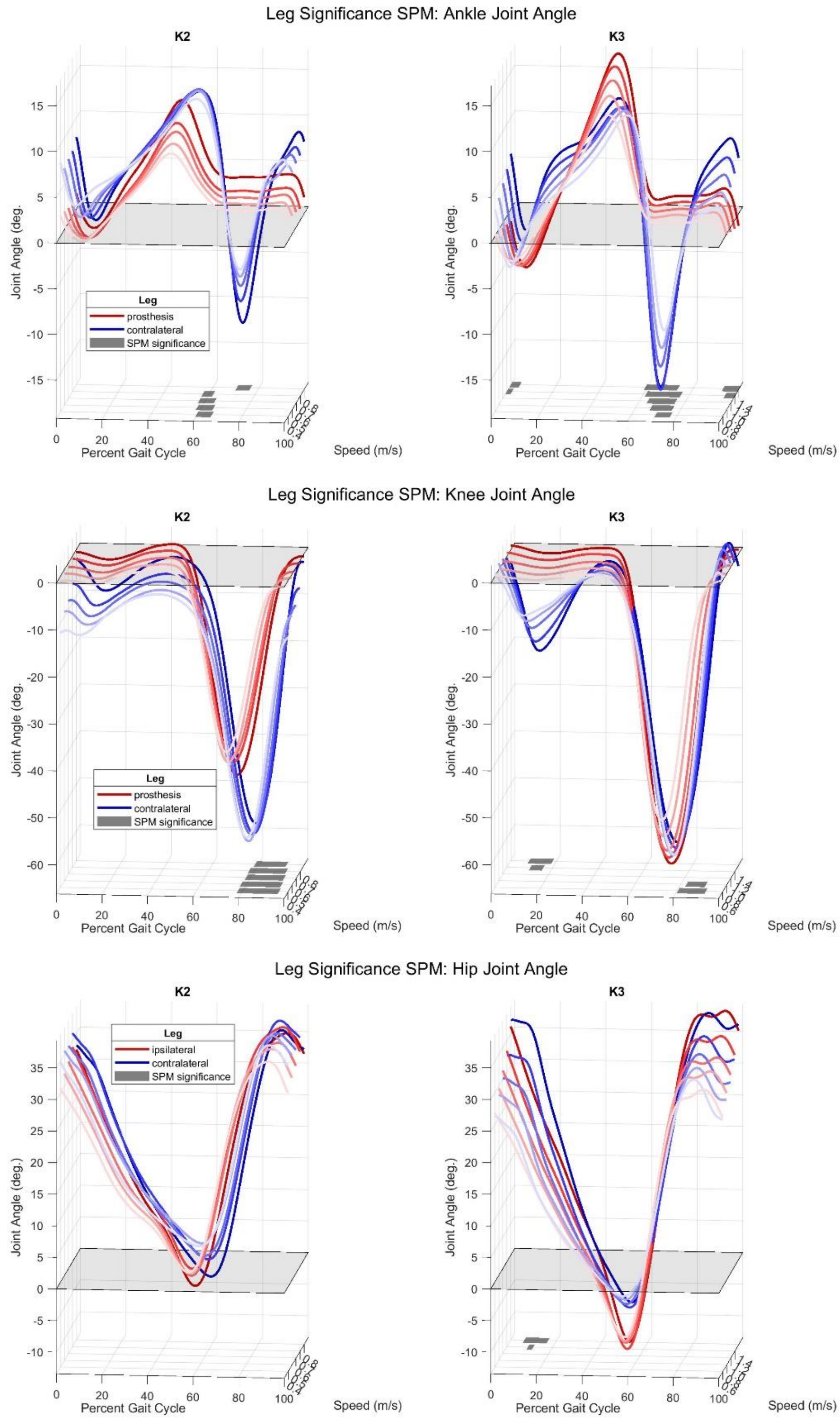

Figure 1: SPM analysis of joint angle over one gait cycle

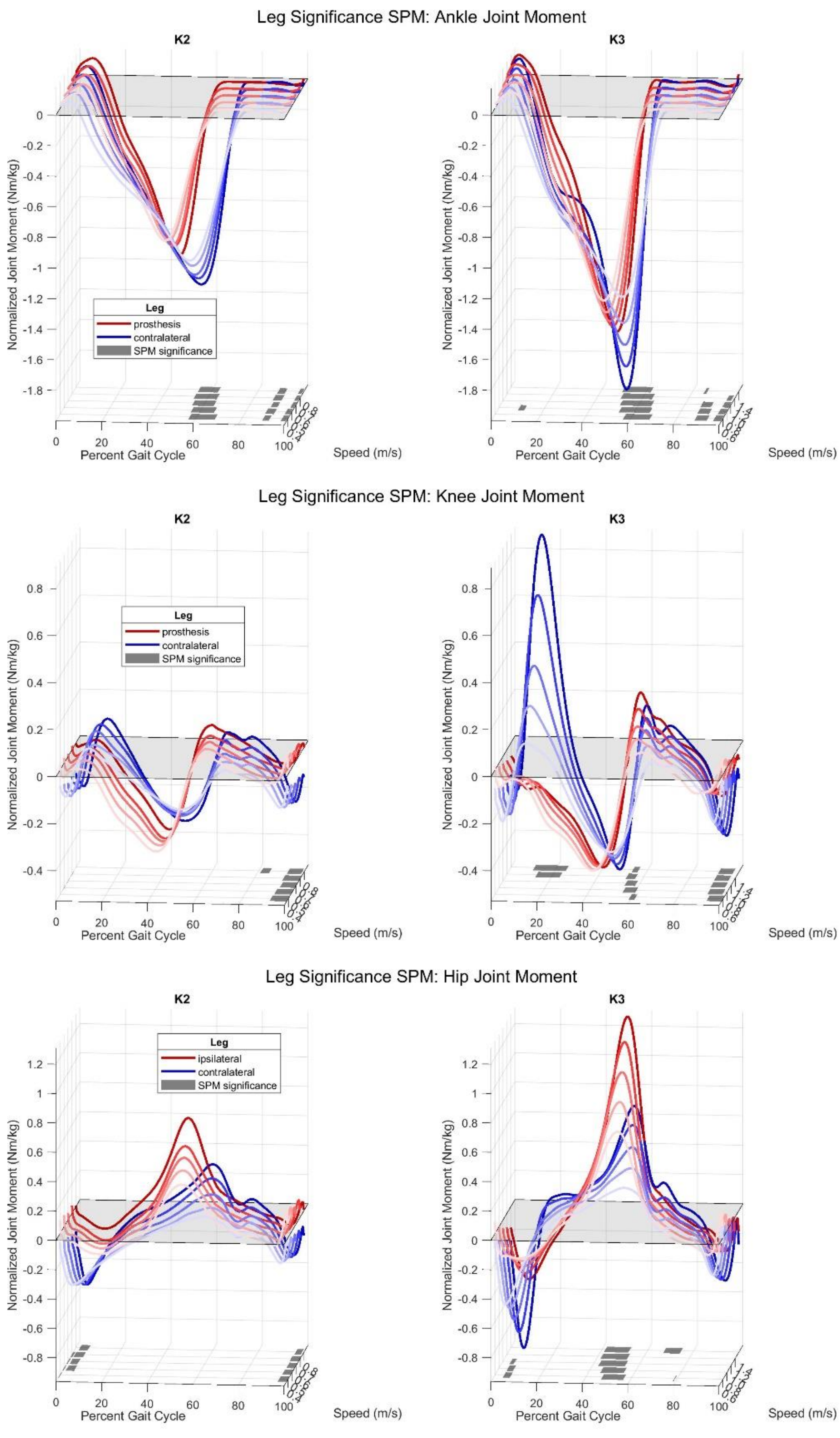

Figure 2: SPM analysis of joint moment over one gait cycle

Ankle Angle

The K3 group had significantly greater contralateral ankle dorsiflexion compared to the prosthetic ankle from ~0% - 5% and 92% - 100% of the gait cycle at their fastest speeds of 1.2 and 1.4 m/s. This region increased in size at the fastest walking speed. The K2 group did not have significant differences between ankle joints in this region. The K2 group had significantly greater contralateral ankle dorsiflexion from ~57% - 67% of the gait cycle except at their fastest speed of 0.8 m/s, where they had lower contralateral ankle plantar flexion from ~59% - 67% of the gait cycle. The K3 group had significantly greater contralateral ankle plantar flexion from ~60% - 75% of the gait cycle, with this region of difference beginning earlier in the gait cycle and increasing in size as speed increased.

Ankle Moment

Both the K2 and K3 groups had significantly lower contralateral ankle plantar flexor moments compared to the prosthetic ankle from ~50% - 70% of the gait cycle, with this region of difference beginning earlier in the gait cycle as speed increased. This region was larger for the K3 group. Both groups had regions of greater ipsilateral ankle dorsiflexor moments from ~87% - 93% of the gait cycle, and greater prosthetic ankle dorsiflexor moments from ~95% - 100% of the gait cycle.

Knee Angle

The K2 group had significantly lower contralateral knee flexion compared to the prosthetic knee from ~78% - 95% of the gait cycle for all walking speeds with the largest regions at the slowest walking speeds. The K3 group only had significantly lower contralateral knee flexion at their two slowest walking speeds of 0.6 and 0.8 m/s, with these regions being much smaller than the K2 group. The K3 group also had significantly lower contralateral knee flexion from ~9% - 18% of the gait cycle at their two fastest walking speeds with this region increasing in size at their fastest walking speed.

Knee Moment

Both the K2 and K3 group had significantly lower contralateral knee flexor moments compared to the ipsilateral knee from ~90% - 100% of the gait cycle with this region staying approximately the same size for all speeds. The K3 group also had significantly lower contralateral knee flexor moments from ~55% - 60% of the gait cycle at all speeds. The K3 group had significantly greater contralateral knee extensor moments from ~10% - 25% of the gait cycle at their two fastest walking speeds with this region increasing in size at their fastest walking speed.

Hip Angle

The K2 groups showed no significant differences in ipsilateral and contralateral hip angle. The K3 group had significantly greater contralateral hip extension compared to the ipsilateral hip from ~8% - 18% of the gait cycle at their two fastest walking speeds. This region of difference was largest at their fastest walking speed.

Hip Moment

The K2 group had significantly lower contralateral hip flexor moments from ~0% - 10% of the gait cycle and ~95% - 100% of the gait cycle at almost all walking speeds with the size staying consistent across speeds. The K3 group had significantly lower contralateral hip flexor moments from ~5% - 10% of the gait cycle at their three slowest walking speeds. The K3 group had significantly greater ipsilateral hip extensor moments from ~42% - 55% of the gait cycle at all speeds with this region beginning earlier in the gait cycle as speed increased. The K3 group also had significantly greater contralateral hip extensor moments from ~68% - 75% of the gait cycle only at their fastest walking speed.

Joint Work

The bar graphs in *Fig. 3* illustrate the average normalized positive and negative joint work over one gait cycle. Red indicates the ipsilateral limb and blue indicates the contralateral limb, with colors getting darker as speed increases. The circles above or below the bars indicate a significant difference between the ipsilateral and contralateral joints at the same speed from the paired t-test.

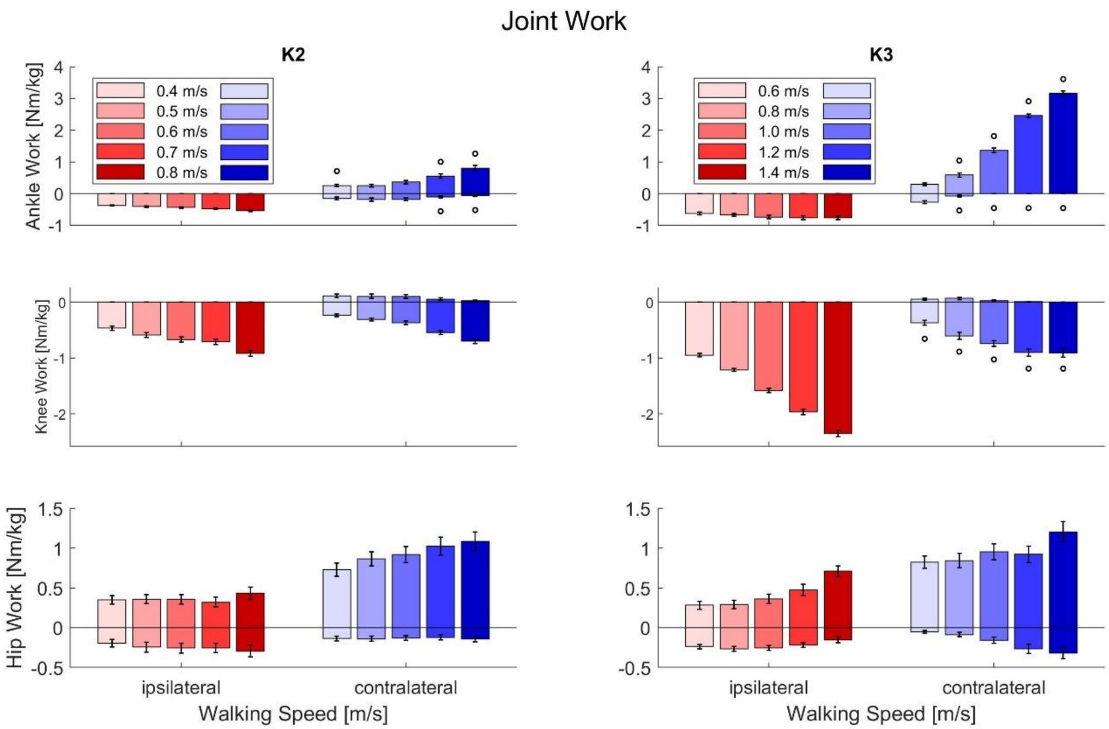

Figure 3: Bar graphs of average normalized joint work over one gait cycle

The prosthetic knee and ankle yielded no positive work for both groups. The contralateral ankle had an increase in positive ankle work and an increase in negative ankle work towards zero as walking speed increased for both groups. The K2 group had five instances of significant difference between the contralateral and prosthetic ankle work, while the K3 group had eight instances of significant difference ($p < 0.05$). Both groups had decreasing negative knee work for both limbs as walking speed increased. The K3 group showed significantly lower prosthetic knee joint work compared to their ipsilateral knee for all walking speeds ($p < 0.05$), while the K2 group showed no significant difference in knee work between limbs. The K2 group had relatively constant positive and negative ipsilateral hip work across speeds. The K3 group showed an increase in positive and negative ipsilateral hip work as walking speed increased. The K3 group also showed an increase in positive and a decrease in negative contralateral hip work as walking speed increased. Neither the K2 nor K3 group showed significant differences in hip joint work between limbs for all walking speeds.

Across Group Comparisons

When comparing the same legs of the K2 and K3 groups while walking at the same steady-state speeds of 0.6 and 0.8 m/s, no significant differences were found in joint angle, moment, or work.

DISCUSSION

The objective of this study was to investigate how the compensatory mechanisms used by individuals with above-knee amputations differ based on mobility classification by examining joint kinematics and kinetics, as well as joint work. Consistent with previous studies and supporting our first hypothesis, the contralateral leg provided significantly more positive work and propulsion than the ipsilateral leg to maintain faster walking speeds for both the K2 and K3 group [17], [29], [30]. However, not only was our second hypothesis that individuals with a higher mobility classification would walk with more symmetry not supported, our data analysis supports the opposite notion – that higher k-level individuals walk with greater asymmetry of joint mechanics than lower k-level individuals. For example, the K3 group had more regions of significant differences between limbs in almost all joint moment and angle trajectories, potentially due to the increased energetic demand to maintain faster walking speeds. The K3 group had increased hip extensor moments in their ipsilateral limb and decreased flexor moments in their contralateral limb, whereas the K2 group only had significantly decreased hip flexor moments in their contralateral limb. Additionally, the K3 group had more significant differences between limbs in both positive and negative joint work at the ankle and knee. Although comparing the same legs of the K2 and K3 groups while walking at the same steady-state speeds of 0.6 and 0.8 m/s yielded no significant differences in joint angle, moment, and work, the K3 group was found to walk with more asymmetry than the K2 group when comparing their ipsilateral and contralateral limbs.

Some of the specific mechanical compensatory mechanisms in the sagittal plane are revealed by investigating the kinematic and kinetic trajectories of the joints at widely-varying walking speeds. Both groups utilized significantly greater contralateral ankle plantar flexor moments towards the end of the stance phase before toe off. This is due to the muscles spanning the ankle injecting energy to maintain faster walking speeds, while the passive prosthetic ankle on the ipsilateral limb can only provide energy from its set stiffness and deflection. The ankle angle trajectories reveal that the K2 group keeps their contralateral ankle loaded in the stance phase longer than their ipsilateral prosthetic ankle. This leads to the K2 group having significantly greater contralateral ankle dorsiflexion from ~57% - 67% of the gait cycle except at their fastest speed. At their fastest speed the K2 group had lower contralateral ankle plantar flexion from ~59% - 67% of the gait cycle.

Aligning with the K2 group's fastest speed, the K3 group keeps both limbs loaded in the stance phase approximately the same percentage of the gait cycle, leading to significantly greater contralateral ankle plantar flexion from ~60% - 75% of the gait cycle. This may be a result of the K2 group requiring less energy to maintain the slower walking speeds they are comfortable with, while their strategy switched to match the K3 group at their fastest walking speed.

The knee joint angle trajectories display the K2 group has significantly lower contralateral knee flexion from ~78% - 95% of the gait cycle for all walking speeds, which is only a strategy used by the K3 group at their two slowest walking speeds. Meanwhile, at their two fastest walking speeds he K3 group has significantly lower contralateral knee flexion from ~9% - 18% of the gait cycle. This aligns with having significantly greater contralateral knee extension moments from ~10% - 25% of the gait cycle at the two fastest speeds. This is likely due to longer stride lengths at faster walking speeds requiring decreased flexion of the contralateral knee and greater knee extension moments just after heel strike to rotate the body over the trunk and provide forward propulsion.

The hip joint angle trajectories support the conclusion about contralateral knee flexion of the K3 group since they also had significantly greater contralateral hip extension from ~8% - 18% of the gait cycle at their two fastest walking speeds. This contralateral hip extension occurs in tandem with the contralateral knee flexion due to longer stride lengths which did not occur in the K2 group. The K3 moment trajectories indicate significantly greater ipsilateral hip extensor moments from ~42% - 55% of the gait cycle, which also does not occur in the K2 group. This increased ipsilateral extensor moment up to the terminal stance phase likely occurs to maintain the increased energetic demands at faster walking speeds.

Mechanically active lower limb prostheses have been the focus of development in recent years in order to provide power to the joints, replacing the functionality of the muscles no longer spanning those joints [31], [32]. A variety of control strategies have been used including microprocessor control [33], variable impedance actuators [34], echo control [35], and gait mode recognition [36], all with the goal of improving the functional performance of individuals with amputation. Previous studies have identified decreased ipsilateral hip extension at the contralateral foot strike, increased pelvic tilt during ipsilateral hip extension, and increased contralateral hip flexion ipsilateral foot strike as significant features of asymmetry during gait cycles of transfemoral amputees [37]. Walking with asymmetry has been shown to increase energetic and metabolic cost, especially for individuals with higher levels of amputation [38]. Studies have also shown symmetry may be the optimal strategy of healthy individuals when energetic cost is estimated as a combined metric of metabolic and mechanical cost [39]. This study has identified regions where compensatory mechanisms are used with much higher temporal resolution of differences in limb mechanics and the effect of an individual's mobility classification and walking speed on their asymmetry. These joint trajectories and regions of asymmetry can be utilized by clinicians for treatment strategies, prostheses selection, and researchers in the development of higher performance active prostheses.

Higher k-level implies higher mobility, however at faster speeds the higher k-level group walked with more kinematic, kinetic, and joint work asymmetry. Although individuals with higher k-level classification generally have more lower-limb strength and resistance to injuries, the more significant regions of asymmetry may lead to more substantial secondary complications. The asymmetric joint trajectories used to walk with passive prostheses motivate a need to design active prostheses controllers to address the needs of individuals with amputation, especially at faster speeds. The findings of this study highlight the common asymmetric features of K2-level and K3-level ambulators and how these compensatory mechanisms change with walking speed. The compensatory mechanisms identified in this paper can be used as a baseline to address the needs of individuals that align with the mobility classifications in this cohort. Regardless of control strategy, adaptive controllers should be designed to improve both kinematic and kinetic symmetry of individuals with amputation.

Appendix: Relevant Subject Information

Subject Code	Age(yrs)	Gender	Mass(kg)	Height (m)	Amputation side	Etiology	Age of Amputation (yrs)	K-Level	Prescribed Prosthesis		Socket Suspension	Training? (#)	Hand-rails?
									Knee	Ankle			
TF01*	26	Male	64.9	1.78	Right	Traumatic	5	K3	Plie FI	AllPro FI	Suction	No	No
TF02**	79	Male	126.1	1.75	Right	Infection	1	K2	C-Leg Obk	Triton Obk	Lanyard	Yes (2)	Yes, All
TF05	72	Male	79.4	1.65	Left	Traumatic	4	K2	C-Leg Obk	Trition Low Profile Obk	Suction	No	No
TF06	60	Male	86.6	1.70	Left	Dysvascular	2	K2	C-Leg Obk	Kinterra FI	Lanyard	Yes (3)	Yes, All
TF07***	49	Male	102.1	1.91	Left	Traumatic	10	K3	C-Leg Obk	Triton Obk	Pin Lock	No	No
TF08	42	Male	95.3	1.85	Right	Traumatic	6	K3	Rheo Os	AllPro FI	Suction	No	No
TF09	65	Male	69.4	1.70	Left	Traumatic	2	K2	C-Leg Obk	Trias Obk	Suction	No	No
TF10	57	Female	58.5	1.65	Left	Traumatic	11	K2	C-Leg Obk	Trias Obk	Suction	No	Yes, All
TF11	51	Male	70.3	1.68	Right	Traumatic	33	K3	C-Leg Obk	Trias Obk	Suction	No	No
TF12	59	Male	99.8	1.83	Left	Traumatic	16	K2	C-Leg Obk	Trias Obk	Lanyard	Yes (1)	No
TF13	61	Male	88.5	1.88	Left	Traumatic	3	K3	Rheo Os	Proflex XC Os	Vacuum	No	Yes, LS
TF14	51	Male	108.9	1.73	Right	Traumatic	3	K2	Genium X3 Obk	Triton Obk	Lanyard	No	Yes, All
TF15	23	Female	68.0	1.75	Right	Traumatic	5	K3	Plie FI	Proflex XC Os	Suction	No	Yes, LS
TF16	36	Male	100.2	1.80	Left	Traumatic	8	K3	C-Leg Obk	AllPro FI	Suction	No	No
TF17	38	Male	104.3	1.91	Left	Traumatic	33	K3	Plie FI	Soleus ClgPk	Suction	No	No
TF18	69	Male	129.3	1.73	Right	Traumatic	50	K2	3R46 Obk	Renegade FI	Suction	No	Yes
TF19	30	Female	59.0	1.60	Left	Traumatic	10	K3	3R80 Obk	AllPro FI	Lanyard	No	No
TF20	59	Male	120.2	1.78	Left	Traumatic	42	K2	C-Leg Obk	Action Obk	Suction	No	Yes

Obk – Ottobock, FI – Freedom Innovations LLC, Os – Ossur, ClgPk – College Park Industries, LS – subject only used handrails on the last speed. *Subject only has 4 walking trials for speeds 0.6, 0.8, 1.0 m/s **Subject was unable to walk at the last speed of 0.8 m/s. ***Subject only has 4 walking trials for speed 0.6 m/s [19].

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